LXXIX

CONSIDERATIONS IN THE MEASUREMENT OF NASAL PATENCY

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The past decade has seen a resurgence of interest in the physiological bases of nasal disease. Accordingly, techniques have been developed to record changes in nasal mucosal temperature and blood flow,[1,2] mucus blanket pH[3] and airway patency.[4-7] Of these variables, the last relates, perhaps best to subjective comfort and has been most widely studied. Experience with methods, designed to reflect nasal airway resistance, has served also to emphasize those features of nasal ventilation which complicate measurement procedures. Since the resulting problem areas, both conceptual and manipulative, limit the validity of data concerning the nasal airway, their analysis and appreciation are a prerequisite for evaluating testing procedures.

Unlike the bronchi, the lumina of which vary with intrathoracic pressure changes, much of each nasal airway acts as a rigid flow channel. (The external nares and nasal vestibules are notable and familiar exceptions.) As such, the same basic principles which govern fluid flow in pipes and wind tunnels will be found applicable to the nasal passages. For simplicity, the following formulation will exclude the significant heat and moisture flux typical of nasal respiration. If the geometry and surface features of these channels are defined, the direction and speed of gas movement are determined by the instantaneous pressure drop existing between the nasopharynx and the atmosphere at the external nares—the transnasal pressure difference (dP). Meas-

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urement of this pressure head has been facilitated by enclosing the nares in an airtight mask.[6,7] If a tube is passed into the mask and another, through the subject's closed lips, into the oropharynx, pressures at these stations may be subtracted manometrically or electronically and their difference recorded, as shown schematically in Figure 1.

Many workers have studied pressure variations occurring at one or the other of these sites, following either the work of Courtade,[8] who measured pressures at the external nares, or applying the principle of "posterior rhinometry" introduced by Speiss.[9] For considerations to be developed subsequently, such isolated pressure recordings have not been wholly satisfactory as indicators of dP or of the airway dimensions it reflects.

It is helpful, in understanding the significance of these pressures, to consider those factors which may contribute to their individually measured values. Atmospheric pressure is exerted at both points even in the absence of air flow, if the pharynx and nares communicate freely. When flow is induced, additional quantities may generate variations in dP. A pressure impulse, proportional to the square of the velocity* (v) of air movement (actually $v^2/2g$), may be sensed at either point. Its measured magnitude will depend upon local air speed and on the precise orientation of the metering tube orifice relative to the direction of the adjacent air flow. This quantity is a form of "dynamic pressure" since it is generated by air motion at the site of measurement. Such "dynamic" effects need not always cause positive pressure recordings, such as occur in Pitot tubes, used to meter the speed of aircraft. Indeed, air flowing perpendicularly to or away from on open oropharyngeal tube, may create a significant stream line convergence with falsely negative pressures at that station. Using the continuity relationship: air velocity equals flow rate divided by cross-sectional area, air velocity ranges in the oropharynx and within appropriate face masks may be estimated. Such calculations do not suggest that significant dynamic pressure effects would be recorded within the range of readily producible nasal air flows (below 0.7 liters/second). By way of confirmation, a large Collins face mask has been used with adult subjects. Except within 1 cm of the nares, measurable changes in pressure recordings could not be produced by changing the position of the sampling tube orifice within the mask. Similarly, if the distance of the pharyngeal tube orifice from the lips

* Velocity is a vector quantity having both magnitude (speed) and direction.

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is constant, but its orientation relative to the supraglottic air stream is varied, no changes in the pressure recordings are discernible. When the distance from the lips to the orifice is gradually shortened, however, small progressive decrements in pharyngeal pressure are seen. Proetz[10] has suggested that large-scale eddying occurs in the pharynx at the level of the fauces, when separation of stream lines from the anterior wall must occur. The finding of diminishing pressure, as the pharyngeal tube is withdrawn into the oral cavity, is consistent with such eddy formation. In practice, where oropharyngeal pressure must be recorded, the length of the oral tube should be maintained constant during sequential measurements.

Most of the instantaneous variations in recorded pressure within the mask and oropharynx are, therefore, independent of the direction of incident air movement at the sampling points. Although thought of as "static," these pressures are not wholly analogous to those that would be recorded if flow was stopped and measurements taken.

The movement of air through any real passage is affected by frictional interaction among the gas molecules themselves and between molecules of air and the surface of the flow channel. Gas viscosity is a measure of the intermolecular friction within air while frictional (or viscous) resistance to flow denotes the loss of measurable energy which occurs when gas molecules are retarded by interaction with the channel walls.[11] This irretrievable loss of momentum with flow creates a significant pressure drop along the nasal airway which increases as nasal obstruction progresses and serves as a relative indicator of nasal patency.

Since the nasal airway is not compliant, i.e. does not change its volume rapidly in response to transmural pressure changes, one might expect that the entire transnasal pressure drop is, indeed, produced by frictional resistance to flow and should be related to changes in cross sectional area. This is certainly the case in uniform straight tubes, especially wide ones. However, a more complex relationship is to be expected in the tenuous and curved nasal passages. The channel dimensions vary strikingly from the choanae (each perhaps 10 mm in width) to portions where only 1 to 2 mm normally separate the opposing walls. Fluid dynamicists are familiar with the additional energy losses which occur at bends in flow channels associated with changes in angular momentum; the development of secondary flow patterns, at these points, is well known. Further losses may occur at "stagnation points" where flow is divided by a mid stream obstruction

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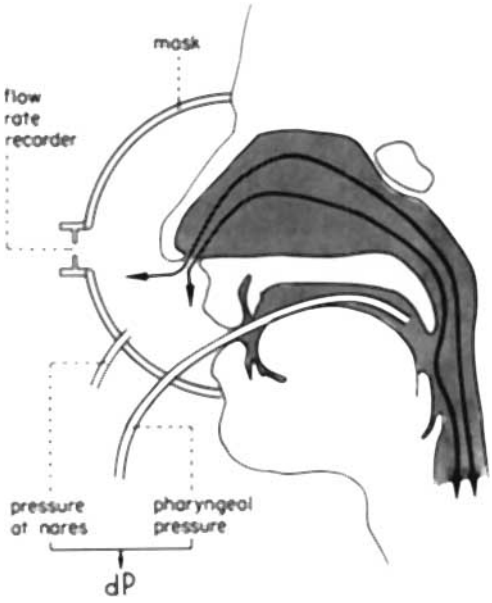

Fig. 1.—Use of a mask and pharyngeal tube to obtain the transnasal pressure drop (dP). The arrows denote approximate paths taken by respired air. Large dynamic pressure effects may be expected at points of flow convergence.

such as the turbinates provide. Unhappily, these losses are readily predictable only for conduits of regular outline. The highly irregular nasal passages, accordingly, present a problem which lends itself more to empirical observation than to facile calculation of these additional energy losses. One can envision the complex and profound effects, on the flow regime, produced by slight engorgement of the venous sinusoids of an inferior turbinate. Significant obstruction to air flow may be produced thereby although much of the nasal passage is unchanged. The resulting increase in transnasal pressure may be readily measured but that portion of the increment which came about due to pure viscous flow resistance cannot, at present, be precisely known. Attempts to relate such an increase in pressure to changes in "average cross sectional area" must reckon with these obvious difficulties.

In addition to compensating for flow-resistive forces, a part of the energy of nasal breathing must overcome the inertia of airway

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gas. The resulting work is proportional to the rate at which gas is accelerated as well as to the mass of the gas involved. It may be recalled that reversal of the direction of air flow between respiratory phrases involves major accelerative forces. Less obvious, perhaps, are the accelerations which gas must undergo in passing narrow sections of a flow passage. Pulmonary physiologists, while aware of inertial forces, properly regard them as small in normal lower airway ventilation.[12] Almost certainly, however, energy necessary to accelerate nasal airway gas can measurably increase the recorded variations in pharyngeal pressure as nasal patency diminishes. Breath-to-breath variations in dP may then be recorded, despite constant frictional airway resistance, if the accelerations applied to airway gas differ markedly in succeeding cycles. For this reason, pressure measurements made a markedly different respiratory rates may not be comparable. Similarly, anxious subjects may make strenuous respiratory efforts with resulting inconsistency in dP although following a metronome. Dependence on modulation of effort by the subject is one factor which seriously limits the value of isolated pressure measurements as comparative indicators of nasal patency. The recording of maximal pressures achievable in expiration, has often been attempted. In addition to accentuating variations due to non-viscous forces, this maneuver may incur an additional defect by producing effort-limiting otalgia.

Unlike energy losses due to viscous resistance to flow, much of the energy required to overcome inertial forces is, at least, potentially recoverable. Its precise measurement would, however, require deceleration of respired gas to rest and determination of the resulting static pressure, a situation not possible in a continuously open system. The use of a mask as a plenum and of mask pressure as a reference point is advantageous and serves to diminish, although not abolish, the effects of inertial forces on dP.

Although dP alone is, at best, an incomplete indicator of nasal patency, transnasal pressure may be meaningfully related to the flow rate (Q) which generates it to provide an airway resistance parameter. To make this measurement, all respired air may be passed through a pneumotachograph. This device employs a system of minute tubes or a fine mesh wire screen across which pressure drops are generated which may be directly related to the rate of incident air flow. Use of a screen of 2-inch diameter affords preservation of a linear response with nasal respiration. Alternatively, a plate orifice has been used to obtain a recording of flow rate.[13]

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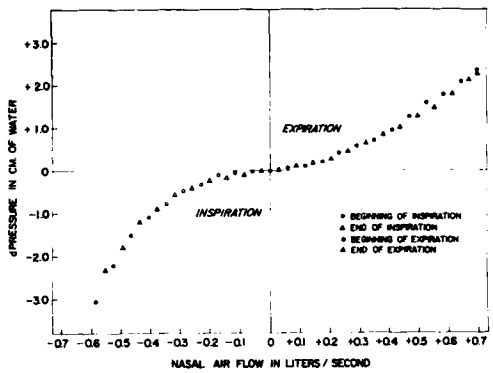

Fig. 2.—The nasal dP-Q relationship for one complete respiratory cycle in a normal subject. Only a short segment of the curve, at lowest flow rates, approaches linearity.

The form of the curve of transnasal pressure versus nasal flow rate, so obtained, has been stressed by Tonndorf[14] using data of Seebohm and Hamilton and of Semerak. Figure 2 shows a plot of such data, on arithmetic axes, for one respiratory cycle of a normal individual. The resulting fiure typically demonstrates only a short portion (at lowest flow rate) that approaches linear form. Since the slope of this curve (delta dP/delta Q) directly reflects nasal airway resistance, the changing value of resistance throughout respiration is readily apparent. This inconstancy in the dP-Q relation associated with nasal respiration requires not only that measured pressures be related to the flow rate with which they are associated, but to be comparable indicators of nasal patency, they must be related to a single specified rate of flow.

Drettner[13] has shown that, for a limited air flow range, such data produce linear plots on logarithmetic axes; a $P = k\ Q^{1.7}$ relationship was postulated as characterizing the nasal P-Q relationship. Figure 3 shows the data of Figure 2 displayed on logarithmetic coordinates. As flow increases, the plot rapidly approximates a straight line with slope 1.7 which may be recognized as the plot of a "conic section": $y = k\ X^{1.7}$.

Figure 3 demonstrates also that, at lower flow rates, the logarithmetic plot is again distinctly curved. This feature suggests that the dP/Q relationship is described by equations having terms in addition

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to the one show above. Furthermore, positive and negative pressure values are frequently recorded at points of zero flow between respiratory phases. Especially when there is significant nasal obstruction or when rapid respiration is tested, plots of dP/Q data for complete respiratory cycles, accordingly, show hysteresis of variable degree. After analyzing curves derived from mechanical models and over 50 subjects, we have concluded that equations of the form $P = k_1 Q^2 + k_2 Q + k_3$ best fit the data points. It is of interest to recall that such a relationship was predicted by Röhrer[15] on purely theoretical grounds. With increasingly large values of Q, dP is progressively less affected by the k_2 and k_3 terms and a $dP = k Q^2$ expression is ultimately approached. Since Drettner examined only flows between 10 and 40 liters per minute, his inference of a more simple $P = k Q^{1.7}$ relationship is understandable. Each half of inspiration and of expiration, then, forms the real half of a parabola symmetrical about the pressure axis. Where hysteresis is encountered, i.e. the dP-Q curve for the first and second half of respiratory phase are not superimposable, the equations for the early and late portions of the corresponding respiratory phase must differ (at least, at k_3).

The optimal means of relating Flow and dP to give a resistance parameter remains a matter of controversy, fostered, in part, by uncertainty concerning the regime of nasal air flow. Following the lead of Röhrer, many workers have assumed that laminar flow normally prevails. In this flow form, a gas or liquid progresses in discrete streamlines which do not exchange significant amounts of mass or eddy momentum; a linear relationship between flow rate and pressure head loss is typical of laminar flow. Viscous resistance to flow, which is calculable as the change in pressure head occurring as the result of a given change in flow rate ($R =$ delta dP/delta Q) is constant with laminar flow and an $R = dP/Q$ relationship validly obtains. Several lines of evidence, however, suggest that nasal airflow is largely nonlaminar and that overt turbulence may often exist. Eddies, which may precede turbulence, are known to occur when laminar flow traverses channels which suddenly enlarge and contract or when obstructions to stream flow are encountered. It has been shown, furthermore, that expired air passing the glottis at moderate velocity already has assumed turbulent characteristics.[16] It would be unexpected for the highly irregular nasal passages to exert a laminarizing effect.

An index of the probability of turbulent flow may be obtained by calculating a dimensionless parameter, the Reynold's Number, for

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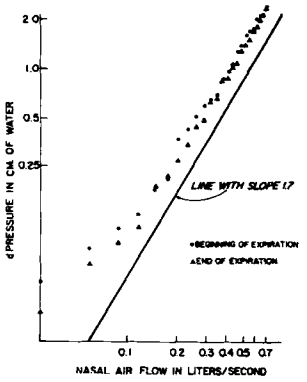

Fig. 3.—Data of Figure 1, for expiration, plotted on logarithmic coordinates. Deviations from linearity are especially prominent at modest flow rates.

given stations in the nasal flow channel. For simple flow in pipes, Reynold's number equals the gas density times the gas velocity times 4 times the profile radius* divided by gas viscosity; values of less than 2000 are generally associated with laminar flow while above 2000, turbulence is the more likely flow regime. Especially for irregular flow channels, however, entrance conditions and surface roughness can change the Reynold's number at which turbulence is established. If measurements derived from anatomical material are used with the above equation and a flow rate of 25 liters/minute is considered, Reynold's number for each choana would be 1335 and for each mucocutaneous junction, 3220. Even without other considerations, these calculations suggest that pure laminar conditions would not be expected at this level of flow. It should be emphasized that sudden transitions from laminar to turbulent flow are seldom seen. Rather, a prolonged period of eddy formation or conditional turbulence, often exists in which dP will be proportional to Q or Q^2 as the flow velocity rapidly alternates in magnitude and direction. This inconstancy of the dp-Q relationship, characteristic of eddy formation, suggests the theoretical advantage of measurements and calculations on flow rates above 0.5 liters/second where turbulence may be expected to be established.

* The profile radius or "Hydraulic mean depth" is a parameter defined as the area of channel section divided by its wetted perimeter and is applicable to non-circular conduits.

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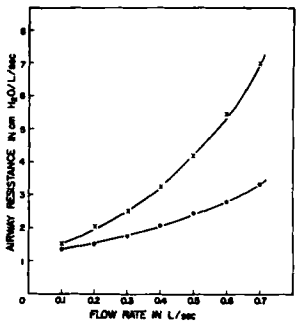

Fig. 4.—Air way resistance for data of Figure 1, in expiration, expressed as dP/V (•——•——•) and as the true resistance, delta dP/delta V (x——x——x). As flow rate rises, the relative error, introduced by the use of dP/V, progressively increases.

The form of the dP-Q curve, previously described, strongly suggests the presence of turbulence. Under this condition, although dp/Q may arbitrarily be employed as a variable, it tends to understate the true resistance (delta dP/delta Q) with which the work of nasal breathing must cope. Long et al.[17] have pointed out that the error associated with a dP/Q approximation, increases as the dP/Q relationship progressively departs from linearity. Figure 4 shows a plot of "resistance" calculated by both of these relationships for the data of Figure 2 obtained in the second half of expiration. As previously mentioned, values for delta dP/delta Q are obtainable as the slope of tangents constructed to the dP-Q curve at the flow values desired. Determination of the slope of such a tangent may be simplified if the dP/Q curve can be projected onto the axes of an oscilloscope. The curve may then be photographed or tangents estimated directly from the luminous image. If dP and Q are synchronously recorded on paper, tangents to both curves may be constructed corresponding to the flow rate at which comparison is to be made. The slope of the line tangent to the pressure curve divided by the slope of the line tangent to the flow curve is then equal to the slope of the line tangent to the dP-Q curve at a chosen flow rate.

Regardless of the resistance parameter chosen, however, the complexity of the nasal passages prevents our calculating back from frictional resistance to the associated dimensions of the nasal flow channels. Under turbulent conditions in simple channels, dP may be expressed

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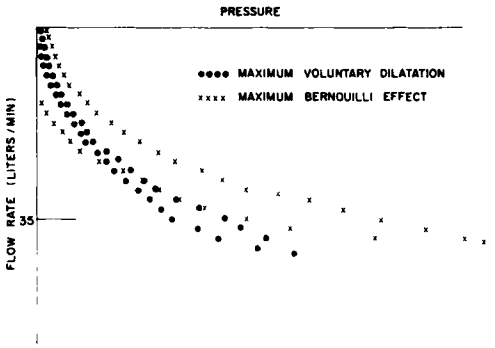

Fig. 5.—Inspiratory dP-V curves made with voluntary augmentation and suppression of nasal alar dilatation. Alar flaring normally offsets the indrawing of the alae resulting from decreased static pressure in the nasal vestibules (Bernouilli effect).

by the equation: $dP = F\frac{v^2}{g}\frac{(L)}{(R)}$. Here, F accounts for the channel roughness and $\frac{L}{R}$ for channel geometry; v is average air velocity, and g the acceleration due to gravity. The applicability of such a relationship to nasal models is currently undergoing study.

Although the internal nasal passages do not exhibit compliance, the mobility of the external nares and nasal vestibules deserve emphasis. Van Dishoeck[18] has shown that inspiratory nasal alar motion is associated with reflex stimulation of its intrinsic musculature and that this effect can be markedly augmented at will. Erratic changes may occur also with psychic stress. Figure 5 shows inspiratory nasal dP-Q curves made with maximal and minimal voluntary alar flare. The variations in resistance resulting from this effect make inspiration ill-suited for precise measurements. However, the first half of expiration may also show distention of the nasal vestibules by the exiting air stream when expiration is rapid. Presumably, air on passing the narrow mucocutaneous external ring (the "ostium internum" of Mink)[19] decelerates in the nasal vestibule. The loss of momentum is attended by a static pressure increase that dilates the pliable nasal alae. The second half of expiration offers a period relatively free of

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such potent sources of error and appears to be an optimal time for measurements.

The induction of mucosal edema and hypersecretion by nasal instrumentation was well described by Mudd et al.[20] and most recent workers have correctly avoided measurement techniques which involve manipulation of the nose. Compression of the nasal mucosa, by the insertion of a tube into the airway may mechanically decrease flow resistance, this effect persisting for a short time after its removal. Thereafter, a more protracted "rebound" mucosal swelling is generally observed. Painful manipulation of the squamous epithelial-lined portions of the nasal passages can also induce mucosal changes; possible effects of less noxious stimuli have not been well studied.

It is probable that procedures which reduce air flow in one or both nostrils can modify the mucous membrane of the corresponding airway. Edema and epithelial changes have been produced in animals whose nares have been sealed;[21] mucosal changes seen in laryngectomized humans who have been deprived of nasal respiration may reflect similar mechanisms.[22] In addition to such long-term effects, decreases in nasal patency have been observed to attend much shorter periods in which nasal ventilation has been suspended.[23]

Nasal vascular volume is, presumably, quite sensitive to changes in pressure occurring in its venous drainage. Relative nasal obstruction induced by recumbency was recognized by Winslow et al.[24] and others have observed mucosal swelling occurring with jugular venous compression. Techniques of nasal measurement that use nasal masks require air tight application to the faces of subjects examined. In achieving this, however, extreme pressure on the nasal bridge and intraorbital areas may induce rapid increases in nasal flow resistance. Transient obstruction of the subcutaneous venous and lymphatic channels in these areas, especially those draining to the anterior facial veins and submaxillary nodes, must be suspected as a causative factor.

SUMMARY

Relative changes in nasal patency may be quantitatively assessed by methods which relate nasal air flow to the associated pressure head across the nasal passages. The resulting data may reflect both viscous losses and dynamic effects relating to gas inertia and to gas velocity at sampling points. Except at the lowest rates of flow, the nasal dP-Q relationship is markedly non-linear; for each half of expiration, a

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$dP = k_1 Q^2 + k_2 Q + k_3$ expression has been found valid. Nasal airway resistance steadily increases with flow rate making sequential determinations comparable only when a common flow rate is used. As nasal patency decreases, the value dP/Q becomes progressively less satisfactory as an estimator of true resistance, delta dP/delta Q. In addition to considerations arising from gas dynamics, measurement procedures must weigh the consequences of nasal manipulation, alar motion, impaired mucosal ventilation and obstruction of nasal vascular outflow.

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